Motion Estimation, Modeling and Compensation

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Introduction

MRI is intrinsically susceptible to motion artifacts; however, for the same reasons MRI offers numerous techniques for measuring motion[1]. High performance MRI systems have increased the speed of data acquisition and thus reduced the susceptibility to unwanted motion artifacts, while increasing the accuracy of motion estimation. We are now at the limit of gradient switching speeds; peripheral nerve stimulation will not allow faster switching times unless some method is used to shield the patient from the stimulating field, so new techniques will have to work within this limitation.

CINE MRI Acquisition

One of the most effective methods for observing motion is the simplest: obtain a movie of the moving object. Even when the image quality of the individual frames is not great, the appreciation of the motion can be quite satisfactory. For example, CINE acquisitions of the heart can show significant wall motion abnormalities that are not appreciated from the individual still frames. Similarly, these movies of the heart can directly demonstrate asynchronous activation of the ventrical[2].

For a motion that is repeated, such as contraction of the heart or blood flow in major vessels, movies in multiple slices can be reconstructed from data obtained from 2-16 consecutive heartbeats during a patient breath-hold. If the patient can hold their diaphragm still for this long, and the ECG trigger works well for each beat, and each of the 2-16 beats is very similar, movies of extremely good quality are produced[3-6]. However, it is not uncommon for at least one of the above constraints to be violated in a sick patient. In this case, reduced data acquisition strategies such as TSENSE[7] or UNFOLD[8;9] can be employed to yield real-time movies[10-12].

Spin Tagging

The principle of "tagging" spins with a saturation pulse was first proposed by Morse and Singer to measure bulk flow[13] . It was demonstrated by Zerhouni et. al. that the same principle could be used to visually mark tissue with tagged magnetization to measure the more complex deformations of the heart[14]. Axel and Dougherty subsequently proposed a very efficient scheme for generating parallel planes of saturation throughout the entire imaging volume[15;16]. Many investigators have proposed refinements and extensions of these methods for generating more complex saturation patterns[17-22]. Extensive efforts have been directed towards validating these patterns for measuring deformation with phantom experiments of both stationary[23;24] and moving samples[25-27].

We can break down the process of motion tracking with spin tagging into three stages: (1) a saturation pattern is placed in the myocardial tissue with spatially selective rf pulses, (2) a sequence of MR images is obtained in which the displacement of the saturation pattern can be observed, (3) the motion of the saturation pattern is used to solve for the motion of the tissue. Consideration of the rate and extent of the motion must be taken into account when designing the appropriate tagging pattern and imaging protocol[27]. Because of the simple nature of the saturation pattern, virtually any imaging sequence can have tagging pulses added to it.

Imaging Tagged Mangetization

Although tagging pulses can be added to any imaging sequence, the imaging sequence must provide good contrast between the tagged and non-tagged tissue[6;28] and provide the temporal and spatial resolution required to track the saturation patterns[27].

Probably the most common sequence used to image the tagging pattern is the segmented k-space GRE or SSFP cardiac gated CINE sequence. The advantage of segmented k-space imaging is that the signal is sampled within a brief readout window with very short TE, (utilizing fractional echoes). This reduces flow artifact and loss of signal due to magnetic susceptibility artifacts. The disadvantage of segmented k-space imaging is the relatively low efficiency of these methods because an rf pulse is required for each line of k-space[29]. Also, the contrast between the tags and the background tissue decreases as both are driven towards the same steady-state with many imaging pulses[6;28;30]. Echo planar imaging yields high efficiency, high tag contrast images; however, single shot EPI images of the heart are extremely difficult to obtain on a consistent basis because of local magnetic field inhomogeneities[31;32]. A compromise of segmented echo-planar imaging[33;34] allows us to tune the readout duration to its optimum. In the heart, the optimal total EPI readout duration is about 10ms. The addition of using TR values of less than 5 ms allows us to perform segmented EPI readouts while maintaining an SSFP type contrast and SNR[5;6;35]. Through the variation of imaging tip angle, the user is able to tune the contrast to be more like bright blood SSFP (with lower tag contrast at later time points in the heart cycle), or darker blood SPGR (with extended tag contrast). The value of the optimal imaging tip angle will depend on the T1 and T2 of the tissue being imaged. If one were tracking displacement of the brain with SSFP tagging, the parameters would be significantly different that those in the heart.

Velocity Encoding Techniques

Measuring tissue motion with velocity encoding was originally proposed by Van Dijk [36] and further developed by Pelc et. al.[37] for CINE acquisitions. In contrast to the tagging methods which image prepared longitudinal magnetization, the velocity encoding methods obtain information about the dynamics of the tissue by phase encoding the velocity of the transverse magnetization shortly before the echo readout. This gives velocity encoding methods an advantage over tagging techniques because they do not suffer from tag pattern fading due to T1 recovery; however, they are more susceptible to motion artifacts. The development of fast switching gradients have reduced this susceptibility significantly, and remarkable datasets of blood flow patterns have been achieved[38]. The trajectory of material points in the tissue of interest can be computed by integrating the three dimensional velocity field, or spatial gradients in the velocity field can be used to derive the strain-rate tensor in soft tissue directly[37;39;40].

Motion Tracking with Velocity Encoding Techniques

Because velocity encoding is a difference technique, at least four acquisitions are required to obtain the three dimensional velocity vector at each image pixel. Velocity encoding of the transverse magnetization is usually accomplished with bipolar flow encoding pulses. To obtain the three dimensional velocity at a specific point, the optimal combination of bipolar pulses is a tetrahedron[41] but deviations from this pattern are possible depending on the circumstances[12]. The different flow encoding acquisitions can be obtained in four contiguous TR periods, or obtained at the same time delay from the ECG in four heartbeats.

Harmonic Phase (HARP) Imaging Methods

In the HARP[42] imaging method a sinusoidal tagging pattern is applied to the myocardium with a simple 1-1 SPAMM[16] tagging pulse. This sinusoid moves the peak of the raw data to a region of k-space centered around the frequency of the sinusoid. If an image is reconstructed from a window of k-space around this peak, the magnitude of the image will be a low resolution picture of the amount of signal in the tagged sinusoid, and the *phase* of the image will show a banding pattern that deforms with the myocardial motion. Relative motion between movie frames of HARP images can be computed from the phase differences, hence motion can be measured without segmenting the position of the underlying tags. The elimination of the tag segmentation step in the analysis can reduce the analysis time by many orders of magnitude. The drawback is that the image is a relatively low resolution representation of the deformation field due to the fact that a window of k-space was used to reconstruct the HARP images. However, the fact that a small region of k-space is acquired means that the imaging can be performed very fast[43]. The same encoding principle can be used to measure longitudinal strain while imaging in the short axis plane[44]. This technique is particularly useful for measuring displacement fields that are spatially smooth.

Displacement Encoding using Stimulated Echoes (DENSE)

DENSE[45] imaging techniques use image phase to encode the net displacement of a pixel over time. Transverse magnetization is position-encoded by a short gradient pulse, stored as longitudinal magnetization during the "mixing time" (denoted TM), and then refocused before image data acquisition by an "unencoding" gradient pulse. The unencoding gradient pulse causes the transverse magnetization to refocus at a phase angle that is linearly proportional to the displacement that occurred during the mixing time. Unfortunately, only ½ of the original transverse magnetization is recovered. However, the DENSE image has displacement estimates on the same spatial scale as the image pixel grid and those displacement estimates do not require image segmentation. These properties make DENSE a very attractive method for quantitative estimates of strain and displacement. Direct visualization of the myocardial deformation is not available, but it can be achieved by driving the motion of "synthetic" tags on the heart with the underlying measured displacement measurements. A number of variants of the DENSE technique have been developed to increase the SNR of the displacement estimates and reduce the effects of T1 recovery[46-50]. One drawback of DENSE is that it is a phase based technique. There are many causes of phase errors in imaging pixels around the heart, such as field

inhomogeneities, chemical shift, motion of the blood, and motion during breathing. Unfortunately, when the value of the phase of a pixel *is* the motion estimate, that estimate is susceptible to error from these many sources.

Modeling Motion

In order to obtain precise quantification of the tissue motion with spatial tagging techniques the position of the tags must be measured with a tag detection algorithm[18;24;51;52], and the three dimensional motions of the tissue computed from the combination of partial displacement information from each tag. A number of approaches can be employed to perform this estimation; most of the work to date has been done in modeling the motion of the heart[53-56]. General motion field reconstruction can be achieved with techniques such as B-splines in space and time[57-60]. Also, alternative methods exist for tracking motion such as optical flow[61], HARP tagging[42], and direct visualization of strain encoded MR[44]. This motion modeling has been investigated extensively in modeling cardiac function[62].

Motion Artifact Reduction

In many applications, patient motion is not the target, it is the problem. For example, high resolution imaging of the coronary arteries requires long acquisition times in order to achieve sufficient signal-to-noise within each small voxel. Unless we are able to increase the polarization of the coronary blood, data acquisition will likely remain in the "minutes" time frame for this application. Motion artifacts can appear due to displacement of the target from view to view; these can be reduced to varying degrees with phase encode view re-ordering methods[63-65]. For certain applications, the motion of the object from view to view can be determined from the imaging data itself[66-70] or from specially crafted navigator pulses[71;72]. Also, there is the potential for correcting image displacement in the spatial domain after full or partial image reconstruction[73;74].

Reference List

1. Ozturk C, Derbyshire JA, McVeigh ER. Estimating motion from MRI data. *Proceedings of the IEEE*. 2003;91:1627-1648.
2. McVeigh ER, Prinzen FW, Wyman BT, Tsitlik JE, Halperin HR, Hunter WC. Imaging asynchronous mechanical activation of the paced heart with tagged MRI. *Magn Reson Med*. 1998;39:507-513.
3. Atkinson DJ, Edelman RR. Cineangiography of the heart in a single breath hold with a segmented turboflash sequence. *Radiology*. 1991;178:357-360.
4. McVeigh ER, Atalar E. Cardiac tagging with breath-hold cine MRI. *Magn Reson Med*. 1992;28:318-327.
5. Herzka DA, Kellman P, Aletras AH, Guttman MA, McVeigh ER. Multishot EPI-SSFP in the heart. *Magn Reson Med*. 2002;47:655-664.
6. Herzka DA, Guttman MA, McVeigh ER. Myocardial tagging with SSFP. *Magn Reson Med*. 2003;49:329-340.
7. Kellman P, McVeigh ER. Ghost artifact cancellation using phased array processing. *Magn Reson Med*. 2001;46:335-343.
8. Madore B, Glover GH, Pelc NJ. Unaliasing by fourier-encoding the overlaps using the temporal dimension (UNFOLD), applied to cardiac imaging and fMRI. *Magn Reson Med*. 1999;42:813-828.
9. Sedarat H, Kerr AB, Pauly JM, Nishimura DG. Partial-FOV reconstruction in dynamic spiral imaging. *Magn Reson Med*. 2000;43:429-439.

10. Guttman MA, Kellman P, Dick AJ, Lederman RJ, McVeigh ER. Real-time accelerated interactive MRI with adaptive TSENSE and UNFOLD. *Magn Reson Med*. 2003;50:315-321.
11. Herzka DA, Derbyshire JA, Kellman P, McVeigh ER. Single heartbeat cardiac tagging for the evaluation of transient phenomena. *Magn Reson Med*. 2005;54:1455-1464.
12. Nezafat R, Kellman P, Derbyshire JA, McVeigh ER. Real-time blood flow imaging using autocalibrated spiral sensitivity encoding. *Magn Reson Med*. 2005;54:1557-1561.
13. Morse OC, Singer JR. Blood velocity measurement in intact subjects. *Science*. 1970;170:440-441.
14. Zerhouni EA, Parish DM, Rogers WJ, Yang A, Shapiro EP. Human heart: tagging with MR imaging--a method for noninvasive assessment of myocardial motion. *Radiology*. 1988;169:59-63.
15. Axel L, Dougherty L. Heart wall motion: improved method of spatial modulation of magnetization for MR imaging. *Radiology*. 1989;172:349-360.
16. Axel L, Dougherty L. MR imaging of motion with spatial modulation of magnetization. *Radiology*. 1989;171:841-845.
17. Bolster BD, Jr., McVeigh ER, Zerhouni EA. Myocardial tagging in polar coordinates with use of striped tags. *Radiology*. 1990;177:769-772.
18. McVeigh ER, Zerhouni EA. Noninvasive measurement of transmural gradients in myocardial strain with magentic resonance imaging. *Radiology*. 1991;180:677-683.
19. McVeigh ER, Bolster BD, Jr. Improved sampling of myocardial motion with variable separation tagging. *Magn Reson Med*. 1998;39:657-661.
20. Mosher TJ, Smith MB. A DANTE tagging sequence for the evaluation of translational sample motion. *Magn Reson Med*. 1990;15:334-339.
21. Pipe JG, Boes JL, Chenevert TL. Method for measuring three-dimensional motion with tagged MR imaging. *Radiology*. 1991;181:591-595.
22. Fischer SE, McKinnon GC, Maier SE, Boesiger P. Improved myocardial tagging contrast. *Magn Reson Med*. 1993;30:191-200.
23. Young AA, Axel L, Dougherty L, Bogen DK, Parenteau CS. Validation of tagging with MR imaging to estimate material deformation. *Radiology*. 1993;188:101-108.
24. Atalar E, McVeigh ER. Optimum tag thickness for the measurement of position with MRI. *IEEE Trans Med Imag*. 1994;13:152-160.
25. Moore CC, Reeder SB, McVeigh ER. Tagged MR imaging in a deforming phantom - photographic validation. *Radiology*. 1994;190:765-769.
26. Salido TB, Hundley WG, Link KM, Epstein FH, Hamilton CA. Effects of phase encode order and segment interpolation methods on the quality and accuracy of myocardial tags during assessment of left ventricular contraction. *J Cardiovasc Magn Reson*. 2002;4:245-254.
27. Kim D, Bove CM, Kramer CM, Epstein FH. Importance of k-space trajectory in echo-planar myocardial tagging at rest and during dobutamine stress. *Magn Reson Med*. 2003;50:813-820.
28. Reeder SB, McVeigh ER. Tag contrast in breath-hold cine cardiac MRI. *Magn Reson Med*. 1994;31:521-525.
29. Reeder SB, McVeigh ER. The effect of high speed gradients on fast gradient echo imaging. *Magn Reson Med*. 1994;32:612-621.
30. Tang C, McVeigh ER, Zerhouni EA. EPI for improvement of myocardial tag contrast: Comparison with SPGR. *Magn Reson Med*. 1995;33:443-447.
31. Atalay MK, Poncelet BP, Kantor HL, Brady TJ, Weisskoff RM. Cardiac susceptibility artifacts arising from the heart-lung interface. *Magn Reson Med*. 2001;45:341-345.
32. Reeder SB, Faranesh AZ, Boxerman JL, McVeigh ER. In vivo measurement of T*2 and field inhomogeneity maps in the human heart at 1.5 T. *Magn Reson Med*. 1998;39:988-998.
33. Reeder SB, Atalar E, Faranesh AZ, McVeigh ER. Multi-echo segmented k-space imaging: an optimized hybrid sequence for ultrafast cardiac imaging. *Magn Reson Med*. 1999;41:375-385.
34. Epstein FH, Wolff SD, Arai AE. Segmented k-space fast cardiac imaging using an echo-train readout. *Magn Reson Med*. 1999;41:609-613.
35. Reeder SB, Herzka DA, McVeigh ER. Signal-to-noise ratio behavior of steady-state free precession. *Magn Reson Med*. 2004;52:123-130.
36. van Dijk P. Direct cardiac imaging of heart wall and blood flow velocity. *J Comput Assis Tomog*. 1984;8:429-436.
37. Pelc NJ, Herfkens RJ, Shimakawa A, Enzmann DR. Phase contrast cine magnetic resonance imaging. [review]. *Magn Reson Q*. 1991;7:229-254.

38. Thompson RB, McVeigh ER. High temporal resolution phase contrast MRI with multiecho acquisitions. *Magn Reson Med*. 2002;47:499-512.
39. Wedeen VJ. Magnetic resonance imaging of myocardial kinematics. technique to detect, localize, and quantify the strain rates of the active human myocardium. *Magn Reson Med*. 1992;27:52-67.
40. Wedeen VJ, Weisskoff RM, Reese TG, Beache GM, Poncelet BP, Rosen BR, Dinsmore R. Motionless movies of myocardial strain-rates using stimulated echoes. *Magn Reson Med*. 1995;33:401-408.
41. Conturo TE, Robinson BH. Analysis of encoding efficiency in MR imaging of velocity magnitude and direction. *Magn Reson Med*. 1992;25:233-247.
42. Osman NF, Kerwin WS, McVeigh ER, Prince JL. Cardiac motion tracking using CINE harmonic phase (HARP) magnetic resonance imaging. *Magn Reson Med*. 1999;42:1048-1060.
43. Sampath S, Derbyshire JA, Atalar E, Osman NF, Prince JL. Real-time imaging of two-dimensional cardiac strain using a harmonic phase magnetic resonance imaging (HARP-MRI) pulse sequence. *Magn Reson Med*. 2003;50:154-163.
44. Osman NF, Sampath S, Atalar E, Prince JL. Imaging longitudinal cardiac strain on short-axis images using strain-encoded MRI. *Magn Reson Med*. 2001;46:324-334.
45. Aletras AH, Ding S, Balaban RS, Wen H. DENSE: displacement encoding with stimulated echoes in cardiac functional MRI. *J Magn Reson*. 1999;137:247-252.
46. Aletras AH, Arai AE. meta-DENSE complex acquisition for reduced intravoxel dephasing. *J Magn Reson*. 2004;169:246-249.
47. Aletras AH, Freidlin RZ, Navon G, Arai AE. AIR-SPAMM: alternative inversion recovery spatial modulation of magnetization for myocardial tagging. *J Magn Reson*. 2004;166:236-245.
48. Epstein FH, Gilson WD. Displacement-encoded cardiac MRI using cosine and sine modulation to eliminate (CANSEL) artifact-generating echoes. *Magn Reson Med*. 2004;52:774-781.
49. Kim D, Epstein FH, Gilson WD, Axel L. Increasing the signal-to-noise ratio in DENSE MRI by combining displacement-encoded echoes. *Magn Reson Med*. 2004;52:188-192.
50. Kim D, Gilson WD, Kramer CM, Epstein FH. Myocardial tissue tracking with two-dimensional cine displacement-encoded MR imaging: development and initial evaluation. *Radiology*. 2004;230:862-871.
51. Guttman M, Prince JL, McVeigh ER. Contour estimation in tagged cardiac magnetic resonance images. *IEEE Trans Med Imag*. 1994;13(1):74-88.
52. Kraitchman D, Young AA, Chang C, Axel L. Semi-automated tracking of myocardial motion in MR tagged images. *IEEE Trans on Medical Imaging*. 1995;14:422-433.
53. Young AA, Axel L. Three-dimensional motion and deformation of the heart wall: estimation with spatial modulation of magnetization--a model- based approach. *Radiology*. 1992;185:241-247.
54. O'Dell WG, Moore CC, Hunter WC, Zerhouni EA, McVeigh ER. Displacement field fitting for calculating 3D myocardial deformations from tagged MR images. *Radiology*. 1995;195:829-835.
55. Denney TS, McVeigh ER. Model-free reconstruction of three-dimensional myocardial strains from planar tagged MR images. *J Magn Reson Imag*. 1997;7:799-810.
56. Declerck J, Denney TS, Ozturk C, O'Dell W, McVeigh ER. Left ventricular motion reconstruction from planar tagged MR images: a comparison. *Phys Med Biol*. 2000;45:1611-1632.
57. Ozturk C, McVeigh ER. Four-dimensional B-spline based motion analysis of tagged MR images: introduction and in vivo validation. *Phys Med Biol*. 2000;45:1683-1702.
58. Amini AA, Chen Y, Elayyadi M, Radeva P. Tag surface reconstruction and tracking of myocardial beads from SPAMM-MRI with parametric B-spline surfaces. *IEEE Trans Med Imaging*. 2001;20:94-103.
59. Amini AA, Chen AY, Curwen RW, Mani VSJ. Coupled B-Snake Grids and Constrained Thin-plate splines for analysis of 2-D tissue deformations from tagged MRI. *IEEE Trans Med Imag*. 1998;17:344-356.
60. Chen CW, Huang TS, Arrott M. Modeling, analysis, and visulatization of left ventricle shape and motion by hierarchical decomposition. *IEEE Trans Patt Anal Machine Intell*. 1994;16(4):342-356.
61. Prince JL, Gupta SN, Osman NF. Bandpass optical flow for tagged MRI. *Med Phys*. 2000;27:108-118.
62. Amini AA, Prince JL. Measurement of cardiac deformations from MRI: Physical and mathematical models. 2001. (ISBN 140200222X) Kluwer Academic Publishers, Dordrecht.
63. Jaffer FA, Wen H, Jezzard P, Balaban RS, Wolff SD. Centric ordering is superior to gradient moment nulling for motion artifact reduction in EPI. *J Magn Reson Imaging*. 1997;7:1122-1131.
64. Nguyen TD, Ding G, Watts R, Wang Y. Optimization of view ordering for motion artifact suppression. *Magn Reson Imaging*. 2001;19:951-957.

65. Huber ME, Hengesbach D, Botnar RM, Kissinger KV, Boesiger P, Manning WJ, Stuber M. Motion artifact reduction and vessel enhancement for free-breathing navigator-gated coronary MRA using 3D k-space reordering. *Magn Reson Med*. 2001;45:645-652.
66. Stehning C, Bornert P, Nehrke K, Eggers H, Stuber M. Free-breathing whole-heart coronary MRA with 3D radial SSFP and self-navigated image reconstruction. *Magn Reson Med*. 2005;54:476-480.
67. Glover GH, Lai S. Self-navigated spiral fMRI: interleaved versus single-shot. *Magn Reson Med*. 1998;39:361-368.
68. Schaffter T, Rasche V, Carlsen IC. Motion compensated projection reconstruction. *Magn Reson Med*. 1999;41:954-963.
69. Larson AC, White RD, Laub G, McVeigh ER, Li D, Simonetti OP. Self-gated cardiac cine MRI. *Magn Reson Med*. 2004;51:93-102.
70. Larson AC, Kellman P, Arai A, Hirsch GA, McVeigh E, Li D, Simonetti OP. Preliminary investigation of respiratory self-gating for free-breathing segmented cine MRI. *Magn Reson Med*. 2005;53:159-168.
71. Moriguchi H, Lewin JS, Duerk JL. Novel interleaved spiral imaging motion correction technique using orbital navigators. *Magn Reson Med*. 2003;50:423-428.
72. Petrie DW, Costa AF, Takahashi A, Yen YF, Drangova M. Optimizing spherical navigator echoes for three-dimensional rigid-body motion detection. *Magn Reson Med*. 2005;53:1080-1087.
73. Stehning C, Bornert P, Nehrke K, Dossel O. Free breathing 3D balanced FFE coronary magnetic resonance angiography with prolonged cardiac acquisition windows and intra-RR motion correction. *Magn Reson Med*. 2005;53:719-723.
74. Kellman P, Larson AC, Hsu LY, Chung YC, Simonetti OP, McVeigh ER, Arai AE. Motion-corrected free-breathing delayed enhancement imaging of myocardial infarction. *Magn Reson Med*. 2005;53:194-200.